

2. Literature Review and Background

2.1 Historical Foundations

The foundations of microdosimetry grew out of the study of radiation effects on living cells. Early attempts at understanding cellular radiation effects recognized that knowledge of the energy distribution at a scale comparable to the structures affected by irradiation was essential. Dessauer [5] and Crowther [6] in the 1920s developed the earliest forms of target theory which identified discrete energy transfers denoted as hits with individual ionization. The neglect of the spatial distribution of these events severely limited the predictive capability of this early work. The late 1940s saw the development of several important concepts, in particular the development of the concept of linear energy transfer (LET). Zirkle [7] in 1952 provided the first definition of LET although his work was closely linked to similar ideas introduced by Gray [8] and Lea [9]. ICRU report no 16 [10] defines LET as a measure of the loss of energy per unit distance along the path of a charged particle. This report also discusses some of the serious limitations of the LET concept in explaining relative biological effectiveness and the differences between radiation types.

Kellerer and Chmelevsky [10], have investigated the effect of these limitations and the ranges and energies over which they are important. There are several reasons for the limitations in the LET concept [11]. Firstly, delta ray energy distribution and its relationship to spatial dose distribution are not adequately considered. Particles with different velocities and charges can have the same LET but it is the particle velocity that largely determines the energy distribution of delta rays. In microscopic volumes, the delta ray distribution may be a significant factor in the spatial distribution of energy, particularly at higher ion energies and small site sizes. Secondly, the limited range of charged particles relative to the finite target structure influences the energy deposition by two possible mechanisms; a change in LET through the site or the possibility of the track beginning or ending within the volume. Such effects are more predominant at lower ion energies and in sites with larger dimensions. Finally, LET, being a non-stochastic average quantity, does not account for the random fluctuations in energy deposition which manifest as clustering of energy deposition and range straggling. The variance due to straggling may exceed the path length variations at high ion energies and small site sizes.

These limitations in LET lead to the formulation of a set of measurable stochastic quantities that provide the fundamental basis for the field of microdosimetry. The two principle quantities are specific energy and lineal energy. These quantities will be defined and discussed in section 2.2. It is the goal of experimental microdosimetry to measure these quantities in well-defined volumes.

Parallel to the theoretical development of radiation interaction with tissue were two main experimental developments [12]. The first main experimental development included studies of visible chromosome aberrations in cells which observed that ionizing radiation could induce exchanges of material between different radiation damaged chromosomes. This work was pioneered by Sax [13] and Lea and Catchside [14] in the 1940s working on *Tradescantia* plant cells followed by Wolff [15] in the late 1950s and Neary [16] and coworkers in the 1960s. Analysis of the size of the affected chromosome structures under different types of ionizing radiation suggested an interaction distance between pairs of damaged chromosomes of between 0.1-1 μm . This compares with a typical cell diameter of 10 μm and a DNA molecule diameter of 0.002 μm .

Clearly, the measurement of radiation quantities in sites of comparable size to the cell or even the chromosome interaction distance are required. Of utility in such measurements was the development in the early 1950s of the low-pressure proportional counter, commonly called the Rossi counter after its pioneer [17]. This was the second main experimental development in microdosimetry and permits measurement of microdosimetric quantities at scales of the order of 1 μm . The operation and limitations of proportional counters will be described in section 2.3.1.

The site concept involves the definition of volumes of interest called sites in which the energy absorbed by ionizing radiation is considered without regard to the microscopic distribution of energy within the site. *Regional microdosimetry* is concerned with the measurement of energy deposition in sites and it is the principal objective of experimental microdosimetry [1]. Intuitively, it would seem likely that a link exists between microdosimetric experimental quantities and the observed effects of radiation on biological cells. The theory of dual radiation action, first proposed by Kellerer and Rossi in 1972 [18], seeks to establish the link between experimental radiation physics and cellular radiobiology. This theory and the applicability of experimental

microdosimetry to radiobiology, radiotherapy, radiation protection, and other applications is discussed briefly in section 2.3.3. Prior to the discussion on applications, the following sections will describe the important concepts of microdosimetry with particular emphasis on experimental microdosimetry. Current experimental methods will be reviewed to facilitate a comparison with silicon based microdosimetry. Past attempts at silicon microdosimetry are thoroughly examined. The main problems associated with silicon based microdosimetry are identified based on this review and the rationale behind utilizing silicon based microdosimetry in medical radiotherapy applications is presented.

2.2 Basic Principles and Microdosimetric Quantities

The formal definitions for the principal microdosimetric quantities are given by the International Commission on Radiation Units and Measurements [11]. Of primary importance is the concept of *lineal energy*, y , which is defined as the quotient of e by $\bar{l}$, where e is the energy imparted to matter in a volume by a single energy deposition event and $\bar{l}$ is the mean chord length in that volume:

$$y = \frac{e}{\bar{l}} \quad (2.1)$$

Lineal energy is commonly presented in units of keV μm^{-1} . The mean chord length in a volume is the mean length of randomly orientated chords in that volume. Various types of randomness exist and the definition of $\bar{l}$ typically refers to m -randomness in which the body is exposed to a uniform isotropic field of infinite straight lines. Chord length distributions and randomness types are further discussed in section 3.3.1 and subsequent sections. According to a theorem by Cauchy, the mean chord length in a convex volume under m -randomness is given by

$$\bar{l} = \frac{4V}{S} \quad (2.2)$$

where V is the volume and S is the surface area of the body.

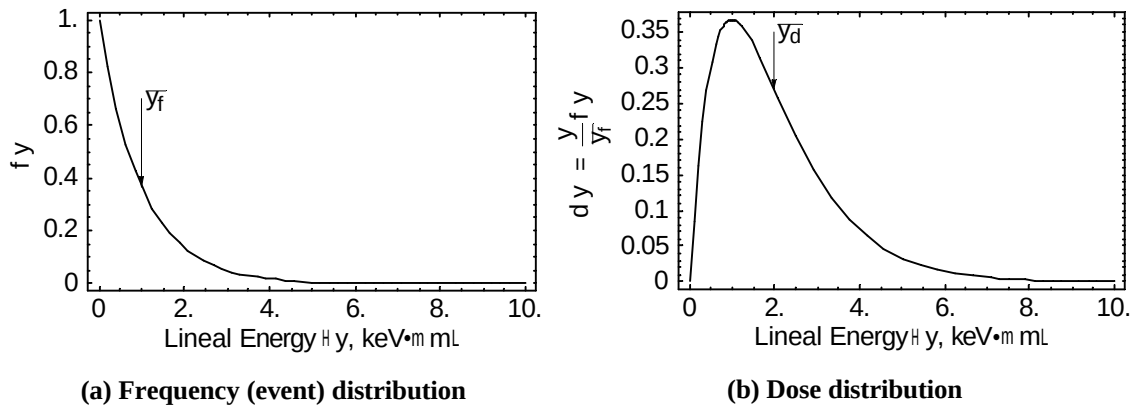
The measured lineal energy is subject to random fluctuations and hence is a stochastic quantity. The random fluctuations are due to all of the previously mentioned limitations in LET, that is, delta ray effects, variations in LET through the volume, and energy and range straggling as well as variations in the chord length through the sensitive volume.

The sources of variation that contribute to the distribution of lineal energy are discussed extensively in section 3.2.

The probability distribution of lineal energy ($f(y)$) is a fundamental function in microdosimetry. An example is shown in Figure 2.1 where a single set of measurements can be displayed as a frequency distribution ($f(y)$ probability of event versus event size) or as a dose distribution ($d(y)$, fraction of energy versus event size). The relationship between $f(y)$ and $d(y)$ is given by:

$$d(y) = \frac{yf(y)}{\bar{y}_F} \quad (2.3)$$

The dose distribution relationship simply reflects the fact that higher lineal energies deposit a higher dose.



The lineal energy was assumed to be exponentially distributed, i.e. $f(y) = \text{Exp}(-y)$, with mean $\bar{y}_F = 1$ keV/ μm . This is not too dissimilar to the measured distributions of single events in a spherical counter of 2 μm simulated tissue diameter irradiated with ^{60}Co γ rays [12]. The lineal energy distributions may be represented as frequency distributions as given by the left-hand diagram or as dose weighted distributions given by the right hand side.

Figure 2.1. Example Single Event Microdosimetric Distributions

The distributions displayed in Figure 2.1, have the following averages defined as

$$\bar{y}_F = \int_0^\infty yf(y) dy \quad (2.4)$$

$$\begin{aligned} \bar{y}_D &= \int_0^\infty yd(y) dy \\ &= \frac{1}{\bar{y}_F} \int_0^\infty y^2 f(y) dy \end{aligned} \quad (2.5)$$

The first moment of $f(y)$ is the frequency mean lineal energy, $\bar{y}_F$, whilst the second moment divided by the first is the dose mean lineal energy, $\bar{y}_D$.

Although not used in this work, another important microdosimetric quantity is specific energy, z , which is the quotient of e by the volume mass m . Similar microdosimetric distributions and averages also apply for this quantity.

Thus, from the single event lineal energy distribution and the site geometry one may calculate all microdosimetric distributions of interest. The representation of these fundamental microdosimetric spectra is traditionally displayed as a log-linear plot with the ordinate multiplied by y such that the area under the curve delimited by two values of y is proportional to the fraction of events (for $f(y)$) or the fraction of dose (for $d(y)$) delivered by events in this range of lineal energy values. This representation accommodates the wide lineal energy range often observed in microdosimetric spectra (from 0.1 keV/ μm to several hundred keV/ μm) but requires further scaling to preserve the dose to area correspondence. Specifically, the ordinate is displayed as $yf(y)$ and $yd(y)$ on a linear-log plot since

$$\int d(y)dy = \int yd(y)d(\ln y) = \ln(10) \int yd(y)d(\log y) \quad (2.6)$$

and this integral is proportional to the dose deposited in the interval dy . Presented in this way, equal areas under different regions of the function $yd(y)$ correspond to equal doses.

As an example, the spectra of Figure 2.1 are represented in standard semi-log form, in Figure 2.2. This form will be used where appropriate in the remainder of this work. A note of caution, when presenting spectra in semi-log form, care must be taken to perform correct normalization given a logarithmically binned histogram, as discussed in Appendix B of ICRU 36 [11]. Replication of Figure 2.1. and Figure 2.2, will assist in verifying the correct algorithms. By definition, the distribution $d(y)$ is normalized to unity:

$$\int_0^\infty d(y)dy = 1 \quad (2.7)$$

This normalization should remain unchanged when plotted versus a logarithmic scale of y using logarithmic binning. Using equation (2.6) and (2.7), the necessary relationship is:

$$\int_0^{\infty} y d(y) d(\ln y) \approx \frac{\ln 10}{B} \sum_{i=0}^{\infty} y_i d(y_i) = 1 \quad (2.8)$$

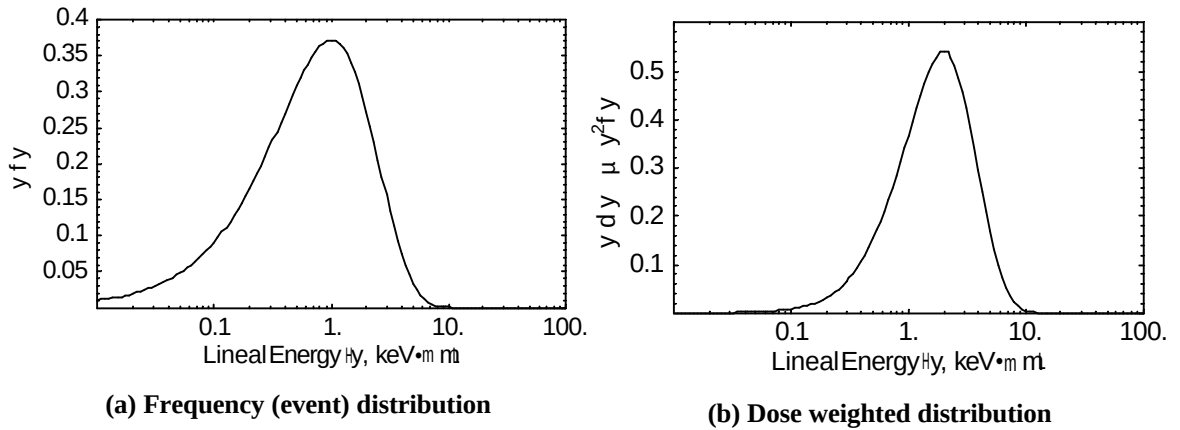
where the logarithmic scale of y is subdivided into B increments per decade such that the i th value of y is:

$$y_i = y_0 10^{i/B} \quad (2.9)$$

The approximation in equation (2.8) arises from the use of the relationship:

$$d(\log y) \approx \Delta(\log y) = \frac{1}{B} \quad (2.10)$$

where the difference between $d(\log y)$ and $\Delta(\log y)$ is negligible provided B is large enough. In all the plots used in this work $B = 40$ which was found to provide sufficient smoothing of the data, adequate display resolution and correct normalization.



Semi-log representations of the spectrum of Figure 2.1. The ordinates have been multiplied by y such that area under the curve is proportional to the fraction of events (or dose) in the areas range.

Figure 2.2. Standard Representation of Microdosimetric Spectrums

2.3 Experimental Methods in Regional Microdosimetry

The following section summarizes current experimental methods in microdosimetry with particular emphasis on their performance capabilities and limitations. The methods discussed include proportional counters, autoradiography, and cloud chambers. The theoretical treatment of proportional counters is complex with limited experimental predictive capability. This review seeks only to elucidate in a largely qualitative manner, the principles of operation and limitations. Note that the use of semiconductor based dosimeters is discussed at length in section 2.3.2.

2.3.1 Proportional Counter Microdosimetry

2.3.1.1 Basic Principles of Proportional Counters

Historically, there have been several difficulties associated with measuring ionization in small micron sized solid devices. These include tissue equivalence problems, small device sizes, and construction difficulties. An alternative approach is to simulate a small tissue volume using a larger gas volume with the requirement that the energy loss of charged particles is identical in both tissue and gas volumes for equivalent trajectories. For a tissue sphere of diameter d_t , density r_t , and mass stopping power $(S/r)_t$ and a gas sphere with parameters d_g , r_g , and $(S/r)_g$, the required condition of equivalent energy loss is[11]:

$$\Delta E_t = (S/r)_t r_t d_t = (S/r)_g r_g d_g = \Delta E_g \quad (2.11)$$

where ΔE_t and ΔE_g are the mean energy losses from the charged particle in tissue and gas. Given a gas with an identical atomic composition to that of tissue and if the mass stopping powers are independent of density then the design criterion becomes:

$$r_t d_t = r_g d_g \quad (2.12)$$

Thus if the ratio of gas to tissue diameter is k_{gt} , then the density of the gas must be reduced from that of tissue by the same factor. The density of the gas may simply be adjusted by changing the gas pressure. As an example, a 2.5 cm diameter sphere filled with propane based tissue equivalent (TE) gas at 17 Torr is equivalent to a 1 μ m diameter sphere of unit density material [19]. The absorbed dose is the same in both spheres but the number of charged particles per unit absorbed dose is multiplied by k_{gt}^2 , the increase in the area cross section.

An important principle in the simulation of microscopic volumes is Fano's theorem [20]. In a medium of constant atomic composition, the fluence of secondary particles is constant if the fluence of primary particles is constant and that under this condition the fluence is independent of the density variations provided the interaction cross-section and stopping powers of the particles are independent of density. The conditions of this theorem must be fulfilled for a counter to correctly simulate tissue. In practice, the counter materials are made tissue equivalent with essentially identical atomic compositions. However, it should be noted that the requirement that mass stopping

powers are independent of density is not always met due to polarization effects in solids [21].

These principles are exploited in the low pressure proportional counter, the most common microdosimetric device. It consists of a spherical or cylindrical gas chamber with a central anode wire traversing the diameter and electrically isolated from the surrounding chamber wall. A voltage is applied between the anode and the conductive wall of the counter such that radiation induced ionization within the chamber results in the drift of ions along the electric field lines. The amount of charge collected is a measure of the number of ion pairs produced. If the average energy (W) required for producing an ion pair is known then the energy deposited within the counter volume may be calculated.

The size of the pulse produced at the anode depends on the applied voltage. At low voltages (a few volts), the ions move relatively slowly and recombination is a significant process. As the voltage increases, the so-called ion chamber region is reached in which almost all ions are collected. At still higher voltages, gas amplification (or avalanche production) occurs in which each ion or electron acquires enough energy to produce secondary and higher generation ions through collisions with gas molecules. In this region of operation, the charge collected is proportional to the energy deposited and the detector is termed a proportional counter.

The amplification or gas gain is a function, among other factors, of the reduced field E/p (where E is the field strength normally expressed in volts/cm and p is the gas pressure expressed in Torr); at sufficiently large values of E/p ionization by electrons can occur [1, 11, 22]. For the simple case of a cylindrical counter the reduced field at a radius r is given by (p163 [23]):

$$\frac{E}{p} = \frac{V}{pr \ln(r_2/r_1)}$$

where V is the applied voltage and r_1 and r_2 are the radii of the central wire and collection volume respectively. Thus, E/p decreases rapidly with r_2 and gas amplification typically only occurs in a very small proportion of the detector within a few wire diameters of the central anode. Consequently, the amount of charge collected is almost independent of the position of the incident ion and the resulting pulse is proportional to the number of primary ions. The shape of the outer wall results in a field

that is more intense at the ends of the collecting wire than in the center. In order to generate a uniform field in the multiplication region, the central anode wire is typically surrounded by an additional electrode shaped as a helix at a potential 80% of the shell voltage (p 89 [1, 11, 22]). A 12.5 mm radius counter at 28 Torr of TE gas simulating a 1 μm radius will typically have an anode radius of 0.0125 mm (1 mil wire) with a helix electrode radius of 1.5mm (p93 [1]).

As the gas pressure decreases (in an attempt to simulate smaller tissue volumes) the amplification region increases such that the counter output ceases to be independent of the spatial distribution of primary ion pairs in the sensitive volume. In order to maintain the amplification region within the confines of the helix electrode, the voltage must be reduced which has the effect of decreasing gain. Conflicting with the requirement of lower gain is that at lower pressures charged particles lose less energy in the counter. Thus, at low pressures the electronic gain must be significantly increased to provide an equal signal to the pulse height analysis system. Rossi (p94 [1]) states a factor of 10 increase in electronic gain from 1 μm simulated diameter down to 0.5 μm . Noise and stability considerations limit the possible electronic gain increases. The generally accepted limit on the simulated diameter due to expansion of the gas multiplication region is around 0.3 μm , although a well defined limit does not exist because it depends on the criterion used for the maximum tolerable spectrum deterioration as well as the applied electric field and gas pressure [11].

Aside from gas pressure decreases, the simulated volume may also be reduced by using smaller electrode diameters. Kliauga [22] has reported reasonable spectra measurements obtained from an ultra-miniature cylindrical counter 0.5 mm in length $\times$ 0.5 mm in diameter operating at 4.2 Torr and simulating a 5 nm diameter unit density volume. The successful operation of such a counter is not well predicted by current theories of electron multiplication in gases that may be more due to the inadequacies of theory as opposed to experimental limitations. More research is required to understand the behavior and possible spectrum quality issues associated with such a counter.

Note, that apart from these design restrictions a more fundamental limit [24, 25] on the minimum site size also exists due to the fact that the unit of measurement is ionizations and that excitations are not detectable. Single ionizations in tissue equivalent gas are equivalent to a mean energy deposition of 30 eV. At site diameters less than about 0.3

μm , errors in the conversion to energy may be quite large due to the small number of ionizations produced by individual low LET tracks with average energy below 30 eV.

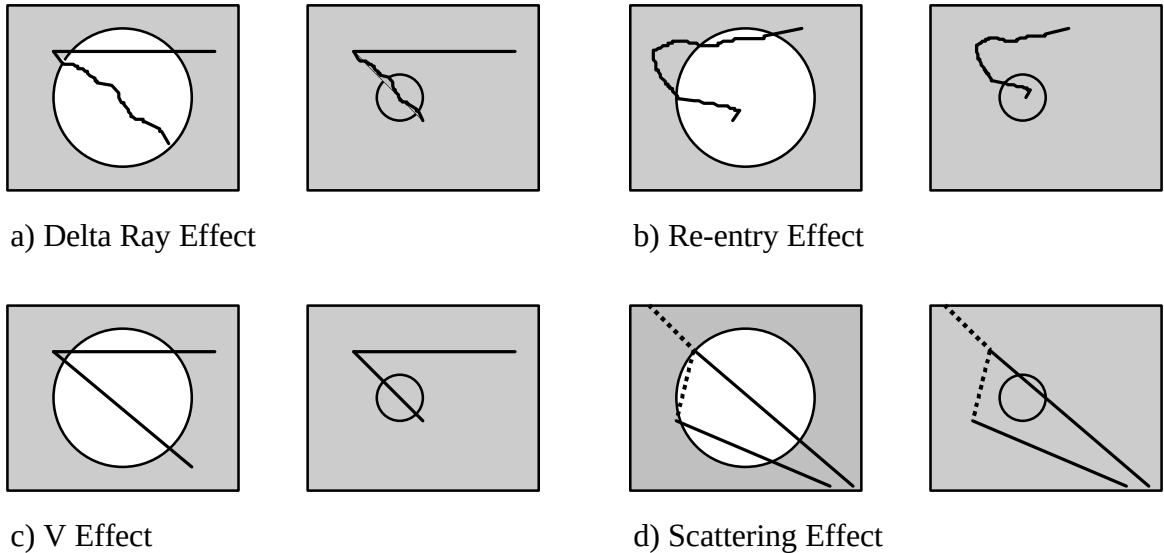
2.3.1.2 Performance of Proportional Counters: Uncertainties and Limitations

The following discusses the important effects that create errors in the experimental measurements performed by proportional counters. Many of the significant sources of error have been listed in ICRU 36 [11] with a notable systematic study also given by Lindborg [26].

Wall Effects: Even if the wall and TE gas have the same atomic composition distortions in the microdosimetric spectra may occur due to density differences and the effect of the non-rectilinear particle traversals. The so-called wall effects are due to the scattering of particles and the production of secondary and tertiary particles. This leads to an increase in energy imparted by superposition of energy deposition events that would not simultaneously occur in a medium of uniform density. Rossi [27] was the first to suggest such effects with a detailed theoretical treatment subsequently given by Kellerer [2, 28]. Wall effects are classified into four types as depicted in Figure 2.3 and described as follows:

1. **Delta-Ray Effect:** A primary particle enters the cavity along with one of its delta rays. However, the distance between the particles is such that only one particle would enter the real tissue volume in the case of uniform density. This effect is most significant for high energy heavy charged particles (significant delta ray production) or for high energy electrons (delta ray and electron may have comparable energies). For heavy charge particles, the energy deposited by the delta ray is much less than the primary particle and hence $\bar{y}_D$ is much less affected than $\bar{y}_F$. Kellerer [2] estimated that for protons of energies above 5 MeV the frequency of double events is about 15% in 1 μm site sizes. For electron energies less than 1 MeV this frequency is less than 6% but the proportion increases at higher energies.
2. **Re-entry Effect:** An electron may re-enter the cavity after it has traversed it due to its tortuous path. The points of entry and re-entry may be far enough apart that the electron would not re-enter the high density microscopic volume. Only low energy electrons (<1 MeV) have sufficient curvature to create this effect. For such electrons, Kellerer [2] estimates about 20% of all energy deposition events will be double events due to the re-entry of the primary electrons.

3. **V-Effect:** The V-effect is similar to the delta ray effect except that the tracks are formed from two heavy charged particles from a non-elastic nuclear interaction. The two tracks make similar contributions and hence unlike the delta ray effect $\bar{y}_D$ as well as $\bar{y}_F$ are substantially overestimated. Oldenburg [29] has performed Monte-Carlo calculations which indicate that the V-effect is insignificant for neutrons up to 6 MeV but increases in importance above 10 MeV.
4. **Scattering Effect:** An uncharged primary particle produces two closely spaced charged particles which both enter the cavity. In the real microscopic region, the two particles may be sufficiently spaced that only one of the particles would enter the sensitive volume. This effect is not well investigated although it may be considered important for neutrons and photons that undergo multiple scattering.



Notes: The meandering tracks represent electron traversal, straight lines are heavier particles, and dashed lines represent uncharged (neutron or photon) primary particles. High-density material is represented by the shaded regions. Each type of wall effect has two diagrams: the left indicates behavior with a walled proportional gas counter and the right diagram illustrates the behavior in equivalent real tissue. In all cases the wall effect leads to an increase in energy imparted by superposition of energy deposition events (left diagrams) which would not occur simultaneously in a volume of uniform density (right diagrams).

Figure 2.3. Diagrams of the four types of wall effect in low pressure proportional counters

The wall-effect has motivated the development of “wall-less” counters of which two different approaches have been taken [11, 30, 31]. The first approach uses electric field lines, circumscribed by small field shaping electrodes, to define the collecting boundary. The main disadvantages of the field defined counter are poor boundary definition due to slight variations in the field and the extreme care required in set up adjustment. A second approach is to delineate the boundary using a fine material grid such that the amount of solid at the boundary is small. Although such a counter has a well defined

boundary and is relatively easy to use it has the disadvantage of not being truly wall-less. Finally, for all wall-less counters particular care should be taken to ensure that the container in which the counter is placed does not disturb radiation equilibrium [2].

Tissue Simulation: Traditionally, proportional counters have been designed to simulate energy deposition in small tissue volumes. Tissue equivalence requires that the mass collision stopping powers of charged particles in the counter and the interaction cross section of the counter materials are identical to tissue. This requirement is fulfilled by utilizing materials with elemental compositions approaching that of standard muscle tissue as specified by ICRU report 44 [32]. Such a requirement conflicts somewhat with practical constraints such as electrical conductivity, toughness, and manufacturability. Varieties of plastics have been developed which closely meet these requirements. The standard plastic formulation for use as a wall material was originally developed by Shonka and designated A-150 [33]. This conducting plastic is a mixture of calcium fluoride, polyethylene (for hydrogen), nylon (for nitrogen) and carbon (as a conductive replacement for oxygen).

The suitability of a tissue equivalent gas is determined by the quality of its elemental tissue composition, gas counting properties and homogeneity with the wall. Of several gas mixtures that meet these requirements two types are predominately used; A methane based mixture developed by Rossi and Failla [34] and a propane based mixture developed by Srdoc [19]. The propane mixture permits higher gas gains and is more homogeneous with A150 whilst the methane based mixture more closely models the ICRU standard.

The quality of the tissue substitute depends on the type and energy of the radiation under investigation. For photons of energy between 50 keV and 5 MeV, A-150 plastic combined with methane based tissue equivalent gas is considered tissue equivalent (p30 [11]). However, below 50 keV differences in total attenuation coefficient are discernible due to the photoelectric effect and above 5 MeV photonuclear reactions occur. For neutrons above 10 MeV the substitution of carbon for oxygen leads to poor tissue equivalence due to differences in non-elastic neutron interactions and inhomogeneity due to carbon-oxygen content differences between the gas and wall. The issue of tissue equivalent materials is revisited in more detail in chapter 5.

Calibration: Generally, the proportional counter is calibrated by irradiating the detector with a source of particles of known energy located near the inside surface of the counter [1, 35]. For high LET radiation, an α -source (e.g. ^{244}Cm , ^{210}Po) is used whilst for low energies, x-ray sources may be employed that emit accurately known K_α radiation that is totally absorbed in the gas by photoelectric absorption. The source spectrum should be fully characterized using a surface barrier detector since thick α sources may exhibit significant spectral spread. For a spherical counter, the maximum of the gaussian pulse-height distribution corresponds to the traversal of a particle across the diameter of the sphere. One may then use the known range-energy relationship for the calibration ion to determine the expected energy deposited. Calibration using an alpha particle source has several sources of uncertainty including source alignment errors, energy degradation of the alpha particle in the source and gain shifts frequently observed between calibration and actual measurements [36, 37].

In the case of neutron or proton irradiation, another convenient method, which does not involve calibration irradiation and removes many of the sources of uncertainty, is based on the phenomenon known as the proton and alpha edge [1, 35]. Microdosimetric spectra under neutron or proton irradiation generally exhibit a step in the spectrum corresponding to the maximum energy deposited by a proton near the end of its range, in the Bragg peak LET region. Specifically, the lineal energy position of the edge, y_{edge} , is given by:

$$\begin{aligned} y_{edge} &= dE_{\max} / \bar{l} \\ &= dE_{\max} / (2/3d) \quad \text{for spherical sensitive volume} \end{aligned} \quad (2.13)$$

where $dE_{\max}$ denotes the maximum energy loss of particles traversing the diameter d . The quantity $dE_{\max}$ is obtained by solving the implicit relation $R(E_i) - R(E_i - dE) = d$ and determining the maximum of dE by varying the initial energy E_i where $R(E)$ is the particle range-energy relationship. The maximum LET of protons in tissue is around 95 keV/ μm (as seen in the Bragg curve) peaking at about 100 keV. Schrewe [37] calculates a proton edge $y_{p-edge} = 136 \text{ keV}/\mu\text{m}$ and alpha edge $y_{a-edge} = 395 \text{ keV}/\mu\text{m}$ for a typical 2 μm spherical TEPC. Pihet et.al. [36] have estimated an uncertainty of 2% each for the estimation of the proton edge position and the stopping power of protons at the edge. In practice, the proton and alpha edges are spread

somewhat due to energy and range straggling and the presence of heavier recoil products that produce events at a lineal energy above and around the edges.

Mean Energy Required per Ion Pair (W): W is the average energy required to generate an electron-ion pair and is frequently needed for calculations related to dose, noise, resolution and sensitivity. For most tissue equivalent gases, W is between 33 and 30 eV (p 88 [1] and [38, 39]). W is known to vary weakly with particle type and energy. For example, Goodman and Coyne [40] have shown that the W in a tissue equivalent methane gas mixture decreases from 32.8 to 31 eV as the energy of the neutrons increases from 0.11 to 20 MeV. W is largely constant for high particle velocities but increases when the particle velocity becomes comparable to the orbital atomic electrons of the gas. Thus, W increases with particle mass (for a given energy) and decreases with particle energy (for a given mass). The uncertainty in W limits the accuracy of the proportional counter in measuring the energy absorbed to around 5% (p 88 [1]). However, one should note that the number of ions generated may well be a better metric for radiobiological efficacy.

Electronics and Sensitivity: The systems required for experimental microdosimetry measurements are shown in the spectroscopy setup of Figure 2.4. The most important component in terms of system noise is the preamplifier. The preamplifier noise is usually specified as the RMS (Root Mean Square) of the number of electrons referred to the preamplifier input. Modern preamplifiers typically have an RMS noise of 400 electrons although by integrating the preamplifier input FET into the counter design, preamplifier noise levels of 100 electrons RMS are achievable [41].

The minimum detection threshold (or Lower Level Discriminator (LLD) setting) of the multi-channel analyzer should generally be set to about 5 times the RMS noise to eliminate the recording of noise counts (p113 [1]). This "rule of thumb" is applicable to both proportional counters and semiconductor detectors. However, an LLD up to 10 times the RMS noise may be required if a low count rate is expected near the LLD energy.

Thus, for a proportional counter the minimum detectable energy E_m is given by:

$$E_m \approx 5 \frac{e_{rms} W}{g} \quad (2.14)$$

where W is the average energy required to generate an electron-ion pair ~ 30 eV, g is the gas avalanche gain and e_{rms} is the system electronic noise referred to the preamplifier input. Typical values are $W = 30$ eV, $g = 10^4$, and $e_{rms} = 400$ giving an $E_m = 6$ eV which is less than the ionization energy for a single ion pair.

However, in order to measure single ions at the anode consideration must also be given to the distribution in charge collected from individual avalanches due to the distances covered between the ionizing collisions of electrons [42]. The spectrum for single electrons is well approximated by a Polya distribution. To ensure all counts are registered the average signal for a single ion-pair should be about 10 times as large as E_m [19]. That is, $10 E_m = W$, which using equation (2.14) means that the gas gain must be 50 times the electronic noise e_{rms} or around 20000 using typical preamplifiers. Such a gas gain is not always achievable, depending on the counter design, and it is desirable to use methods mentioned previously to reduce preamplifier noise. For a given design, gas gain is varied by adjusting the high voltage (500-5000 V) supply. In summary, in order to measure the collection of single ions at the anode, the preamplifier noise should be as low as possible and the gas gain should be high.

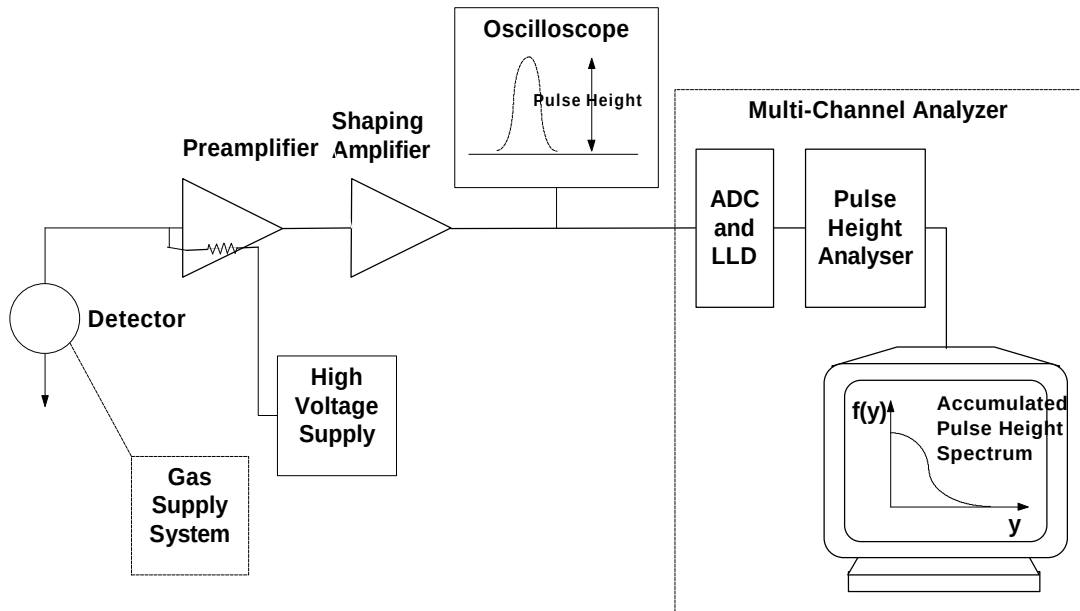


Figure 2.4. Setup required for experimental microdosimetry. With a proportional counter the gas system and high voltage supply are used. For a semiconductor microdosimeter detector the supply may often be low voltage and the gas system is not required. (LLD = lower level discriminator, ADC = Analog to Digital converter)

To optimize noise, a shaping amplifier is included in the spectroscopy experimental setup of Figure 2.4. A time constant of 0.1 μs is often employed for proportional counters which imposes a maximum count rate of 10^4 - 10^5 cps for most applications [1]. Noise optimization and shaping amplifier considerations are discussed in detail in chapter 4. The output of the shaping amplifier should be monitored with an oscilloscope to ensure correct pulse shaping (pole-zero adjustment) and no evidence of signal pile-up. The acquisition of the entire lineal energy spectra using the multi-channel analyzer may require two separate acquisitions with different amplifier gain settings due to the large dynamic range of the energy deposited. The two spectra may then simply be merged.

Energy Resolution: Given microdosimetric events of equal magnitude, the proportional counter will measure a range of energies deposited due to various statistical fluctuations. The resolution of a detector is defined as:

$$R_{FWHM}(\%) = 2.35 \frac{s_E}{\bar{E}} 100 \quad (2.15)$$

where $s_E / \bar{E}$ is the relative standard deviation of the distribution of energy collected. Two classes of fluctuations contribute to the distribution:

1. *Theoretical resolution, R_{thpc} :* Two inherent factors which are largely unavoidable in proportional counter operation contribute to the ultimate theoretical resolution. Firstly, the number of ions produced when energy T is absorbed varies about a mean T/W according to a relationship proposed by Fano [43]. The relative variance V_f is given by:

$$V_f = \frac{W}{T} F \quad (2.16)$$

where F is the Fano factor which is typically 0.3 for proportional counters (p 89 of [1]). Secondly, the previously mentioned statistics of gas multiplication results in another contribution with a relative variance V_m given by:

$$V_m = \frac{W}{T} m \quad (2.17)$$

where m is a multiplication factor which is typically 0.6 for proportional counters (p 96 of [1]). Adding the relative variances gives the theoretical resolution, R_{thpc} :

$$R_{thpc}(\%) = 2.35 \sqrt{\frac{W}{T} [F + m]} 100 \quad (2.18)$$

2. *Instrumentation Resolution, R_{instpc}* : Resolution is affected by imperfections in the counter design including preamplifier noise and variations in counter voltage, gas pressure and electric field uniformity. These contributions are typically less than 10% of the theoretical resolution. Note that the preamplifier noise, e_{rms} , must be corrected for the gas gain, g , to determine the contribution to resolution. Denoting non-preamplifier instrument resolution as R_{rest} , the instrumentation resolution may be expressed as:

$$R_{instpc}(\%) = 2.35 \sqrt{\left(\frac{W}{T} \frac{e_{rms}}{g} \right)^2 + R_{rest}^2} 100 \quad (2.19)$$

Combining equations (2.18) and (2.19) gives the estimated resolution R_{pc} :

$$\begin{aligned} R_{pc} &= \sqrt{R_{thpc}^2 + R_{instpc}^2} \\ &= 2.35 \sqrt{\frac{W}{T} [F + m] + \left(\frac{W}{T} \frac{e_{rms}}{g} \right)^2 + R_{rest}^2} 100 \end{aligned} \quad (2.20)$$

In summary, the dominant factor in the resolution is the gas multiplication variation. The instrumentation resolution is a minor contribution.

2.3.2 Semiconductor (Silicon) Microdosimetry

2.3.2.1 Basic Principles of Semiconductor Detectors

There are many references devoted to the physics of semiconductors. The following section is based on books by Sze [44], and Knoll [23] with the specific goal of providing a brief and basic framework for understanding the operation, capabilities and characteristics of silicon based radiation detectors. Note that the most common semiconductor material for room temperature radiation detection is silicon so the ensuing discussion will use silicon in all examples.

The usefulness of semiconductors as circuit elements and for radiation measurement stems from the special properties created at a junction where n and p type semiconductors are brought into good thermodynamic contact. To ensure excellent contact, the junction is created by ion-implantation or diffusion of appropriate dopants in a single crystal. These techniques and subsequent processing can create a variety of

doping profiles that affect the device characteristics. A useful approximation in many applications is the abrupt junction, which will be reviewed briefly.

Consider the n^+-p junction shown in Figure 2.5 where the type change at the junction is taken to be a step function from a heavily doped n region to a lightly doped p region. The Fermi level (E_F) in the energy band diagram reflects the high concentration of electrons in the n^+ region and the high concentration of holes in the p region. When the junction is first formed large carrier concentration gradients at the junction cause carrier diffusion. Holes from the p -side diffuse into the n -side leaving behind uncompensated negative acceptor ions (N_A^-) and similarly electrons from the n -side diffuse into the p -side leaving behind uncompensated positive donor ions (N_D^+). Note that the acceptors and donors are fixed in the semiconductor lattice while the holes and electrons are mobile. The result of this movement is the formation of a negative space charge on the p -side and a positive space charge on the n -side. The space charge region creates an electric field directed from the n to the p -side that results in a carrier drift current opposing the diffusion currents. Thus, at thermal equilibrium (i.e. no bias applied, steady state condition with no external excitation) the net current flow across the junction is zero.

The space charge region is also called the depletion region due to the very low mobile carrier density. This region acts, like a high resistivity parallel-plate ionization chamber making it feasible to use for radiation detection. The electric field separates electron-hole pairs formed by ionizing radiation and collects the carriers via drift. The rapid movement of carriers in the electric field generates a current pulse at the device electrodes. By applying, a reverse bias to the n^+-p junction, the charge collection capabilities of the device may be improved. The presence of an external electric field in the same direction as the depletion region field (i.e. reverse bias) will cause additional majority carriers to vacate the junction region leaving more uncompensated impurities and thus expanding the depletion region. This effect is seen in Figure 2.5 where the space charge distribution, electric field distribution, potential distribution and band diagram are depicted for zero and 10 V reverse bias. The device selected has parameters similar to the prototype silicon microdosimeter used in this work and described in chapter 4.

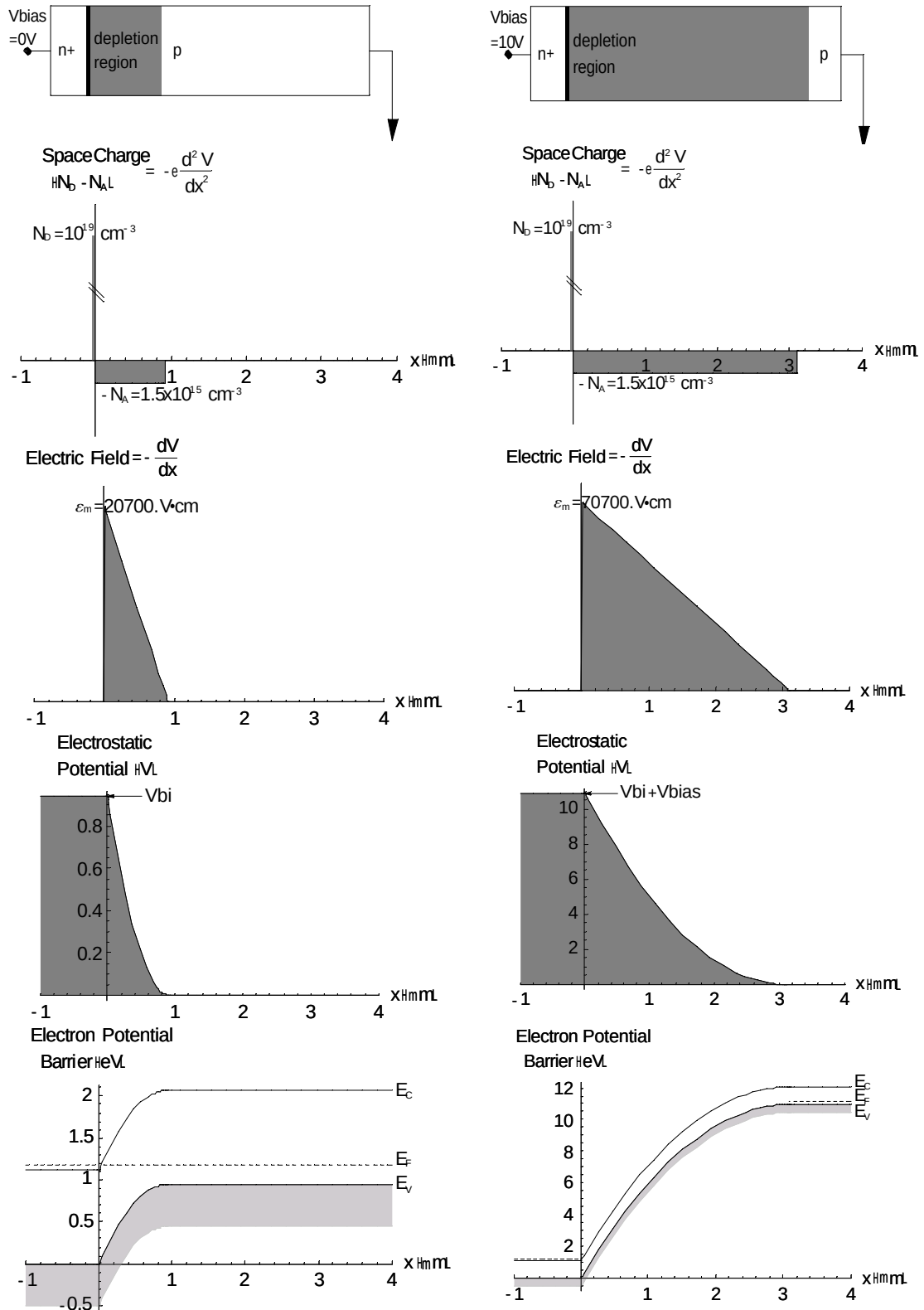


Figure 2.5. A schematic diagram of an abrupt p-n junction including the space charge density, electric field intensity, electrostatic potential and band diagram. The familiar band diagram (showing Fermi energy E_F , conduction E_C and valence bands E_V) may be thought of as the electron potential barrier. The left side shows plots for thermal equilibrium conditions with no bias voltage applied whilst the right hand side shows the same plots for a bias voltage of 10 V. The values selected to construct these graphs correspond to the silicon microdosimeter prototype (p doping $N_A = 1.5 \times 10^{15} \text{ cm}^{-3}$, n doping $N_D = 2 \times 10^{20} \text{ cm}^{-3}$, effective density of states in valence band $N_V = 1.04 \times 10^{19} \text{ cm}^{-3}$, effective density of states in conduction band $N_C = 2.8 \times 10^{19} \text{ cm}^{-3}$, $T = 300 \text{ K}$).

The equations used in the figures apply to a one-sided abrupt junction and are readily available in many texts such as Sze [44]. The key starting equation is Poisson's equation defined for the one-dimensional case as:

$$\frac{d^2V}{dx^2} = -\frac{r}{e_s} \quad (2.21)$$

where V is the electrostatic potential, e_s is the semiconductor dielectric permittivity and the r is the space charge density given by the sum of the charge carrier densities and the ionized impurity concentrations. That is,

$$r = q(p - n + N_D^+ - N_A^-) \quad (2.22)$$

In the depletion region, $r = q(N_D^+ - N_A^-)$ since there are no free carriers. Equation (2.21) may be solved for the electric field intensity and electrostatic potential given appropriate boundary conditions. Some useful results are now presented for the depletion region width, built in potential and maximum electric field intensity.

The depletion region width W [44] for a one sided abrupt junction is:

$$W = \sqrt{\frac{2e_s(V_{bi} - V)}{qN_B}} \quad (2.23)$$

where V is the reverse bias, V_{bi} is the built-in potential, q is the electron charge and N_B is the substrate doping concentration ($= N_A$ in our example).

The built-in potential V_{bi} is given by [44]:

$$V_{bi} = \frac{kT}{q} \ln\left(\frac{N_A N_D}{n_i^2}\right) \quad (2.24)$$

where k is Boltzmann's constant ($= 1.38066 \times 10^{-23}$ J/K), T = temperature (K), q = electron charge, N_A is the acceptor (p) doping concentration and N_D is the donor (n) doping concentration.

The maximum electric field strength e_{max} is given by [44]:

$$e_{max} = \frac{qN_B W}{e_s} \quad (2.25)$$

Equations (2.21) through to (2.25) were applied in the construction of Figure 2.5 and will prove useful at various junctures in this work.

2.3.2.2 Performance of Silicon Microdosimeter

Many performance issues are discussed throughout this work. Given these detailed discussions, this section will summarize the performance to facilitate a comparison with the proportional counter provided in section 2.3.4.

Sensitive Volume Definition: In most attempts at silicon microdosimetry, particularly using existing commercial memory devices, the sensitive volume is poorly defined. This is an important problem in silicon microdosimetry that will be addressed in this work. Charge collection does not simply occur via drift in the depletion region. Diffusion from outside of the depletion region may be a significant contribution to charge collection. Further complicating matters is the funneling phenomenon associated with depletion region collapse following the injection of high carrier concentrations in the device. Clearly, the generation of ion tracks and evolution of the charge collection process are complex phenomenon and will be described in detail in section 6.1.

Radiation Hardness: High-energy radiation produces defect complexes in semiconductor materials which reduce minority carrier lifetime, change majority carrier density and reduce mobility. All of these effects reduce charge collection efficiency effectively altering the sensitive volume size. Radiation damage effects on silicon devices is discussed in detail in section 6.3.

Tissue Equivalence: The difficult issue of tissue equivalence in silicon microdosimeters is the subject of chapter 5. Silicon microdosimeters require a tissue equivalent converter on top of the device. Ideally, detected interactions should arise exclusively from events originating in the medium surrounding the detector (referred to for the remainder of this work as the *converter*). Therefore, the device should have a small overlayer thickness and the sensitive volume should be as small as possible to enhance crosser type particles and reduce direct nuclear interactions with silicon. In chapter 5 it will be shown that, assuming identical secondary charged particle spectra, the microdosimetric spectrum will be closely equivalent in a silicon volume and a tissue volume if the dimensions of the silicon volume are scaled by a factor of 0.63. The reader is referred to chapter 5 for more detailed analysis of tissue equivalence issues.

Mean Energy Required per Electron-Hole Pair (W): At room temperature the mean energy per electron-hole pair is 3.62 eV based on alpha particle measurements (p 348 [23]). This is about 10 times lower than for the proportional counter with positive implications for resolution as will be discussed below. The value of W varies with particle type and energy in a similar manner to TE gases. Differences as large as 2.2% have been observed between proton and alpha particles with smaller differences for other light ions and electrons [45]. The value of W is significantly higher for heavy ions and fission fragments than for alpha particle excitation leading to a lower than anticipated energy collected. This effect arises because as the velocity of heavy ions decreases the probability of losing energy via nuclear collisions rather than electronic collisions becomes significant. Wilkins (p 374 [23] from [46]) demonstrates, by irradiating a silicon surface barrier detector with (He, C, S, Ni, Ag, Au and U) ions, that pulse height versus ion energy does not deviate from linearity when using the He, C and S ions and pulse height deficits are seen for the remaining heavier ions. In this study we will mostly be concerned with ions of atomic weight less than S and $W = 3.62$ eV is a reasonable assumption.

Electronics, Calibration and Sensitivity: The system required for silicon microdosimetry measurements is similar to the proportional gas counter as shown in Figure 2.4. Important exceptions are the removal of the gas system and the need for only a low voltage supply (e.g. battery operated) in low resistivity silicon microdosimeters. System calibration using known energy sources is easy and similar to the proportional counter.

Again, the most important component in terms of system noise is the preamplifier. In fact, the influence of the preamplifier noise is more significant in the silicon microdosimeter because the device does not have any internal (or gas) gain. Thus, for a silicon microdosimeter the minimum detectable energy E_m is given by equation (2.14) with $g=1$:

$$E_m \approx 5e_{rms} W \quad (2.26)$$

For the silicon prototype microdosimeter used in this work $e_{rms} \sim 600$ electrons giving $E_m = 10.8$ keV.

All other electronic considerations discussed previously for the proportional counter apply to the silicon microdosimeter since they share the same amplifier and MCA

components. Noise optimization and shaping amplifier considerations for the silicon microdosimeter will be discussed in detail in chapter 4.

Energy Resolution: The energy resolution of the silicon microdosimeter, like the proportional counter, may be divided into two classes of fluctuations:

1. *Theoretical resolution, R_{thpc} :* The ultimate theoretical resolution for the silicon device is better than the proportional counter for two reasons. Firstly, there is only one component given by the Fano relationship (equation (2.16)) and secondly the Fano factor is smaller for the silicon detector with a typical value of 0.1 employed (p340 [23]). Thus the theoretical resolution, R_{thsi} , is given by:

$$R_{thsi}(\%) = 2.35 \sqrt{\frac{W}{T} F} 100 \quad (2.27)$$

2. *Instrumentation Resolution, R_{instpc} :* The primary component defining instrumentation resolution is preamplifier noise. This is a more significant contribution to overall resolution than for the proportional counter since e_{rms} , is not divided by a gas gain factor in the expression for resolution. Thus, the instrumentation resolution may be expressed as:

$$R_{instsi}(\%) = 2.35 \frac{W}{T} e_{rms} 100 \quad (2.28)$$

Combining equations (2.27) and (2.28) gives the estimated resolution R_{si} :

$$\begin{aligned} R_{si} &= \sqrt{R_{thsi}^2 + R_{instsi}^2} \\ &= 2.35 \sqrt{\frac{W}{T} F + \left(\frac{W}{T} e_{rms} \right)^2} 100 \end{aligned} \quad (2.29)$$

At low energies (T) the preamplifier noise dominants whilst at high energies the Fano factor defines the resolution. The sensitivity and resolution of the silicon detector will be compared further in section 2.3.4.

2.3.2.3 Review of Semiconductors in Microdosimetry

Research on the development of semiconductor based microdosimetry may be classified into two general device types; *single* p-n junction diodes and p-n junction *arrays* based on commercial memory or charge coupled devices:

1) Single Diode Devices: The first (and most detailed to date) comparison of microdosimetric measurements between a spherical proportional counter and a single junction solid-state detector was made by Dicello [47] in 1980. Microdosimetric spectra for range-modulated negative pion beams (momentum=167 MeV/c) were made using a 1.3 cm spherical proportional counter with a simulated diameter of 2 μm and a 7 μm thick, 0.1 cm radius lithium drifted silicon detector.

Pions, as heavy charged particles, exhibit a Bragg-type depth dose distribution which is enhanced around the Bragg peak by the process of star formation arising from nuclear absorption of the pions and subsequent nuclear fragmentation. The secondary products from pion star formation consist of an isotropic emission of light and heavy ions from the fragmented nucleus along with neutrons and γ -rays. Experimental microdosimetry is a useful tool for analyzing the complex secondary particle distribution in pion beams.

For this type of beam, Dicello [47] found significant differences between the microdosimetric spectra obtained by the two detectors. Monte Carlo calculations demonstrated that the differences were largely attributable to the relatively large sensitive thickness of the silicon detector, that permitted a significant fraction of heavy ion stoppers, and the large diameter that permits large chord lengths. Events at large chord lengths although improbable deposit a high energy and may contribute substantially to the total dose. Moreover, low LET particles such as protons may erroneously produce events that contribute to the high lineal energy component of dose. Despite these difficulties, Dicello [47] acknowledged the potential for silicon microdosimetry due to its high spatial resolution, in-vivo capability and pile-up robustness.

The silicon device is advantageous in beams with high instantaneous count rates since the small area of the silicon detector and the higher noise cut-off reduce pile-up problems. The primary proton beam had a time structure of 120 pulses per second and a duty factor of 7.5%. Dicello [47] found the silicon detector was capable of measurements at full therapeutic beam intensity whilst the proportional counter measurements were complicated by a need to reduce beam intensity.

Later work by Dicello in 1986 [48, 49] focussed on the application of experimental microdosimetry to single event upset studies in silicon. Dicello designed a silicon walled proportional counter to study the energy deposition of pions and muons in

simulated 1 μm silicon site sizes. This work was directed towards the study of cosmic radiation induced single event upset since pions and muons are present in the atmospheric cosmic radiation spectrum

Other attempts at creating silicon microdosimeters using single diodes include the work of Orlic in 1989 [50] and later Kadachi in 1994 and onwards [51, 52]. Both efforts were directed towards radiation protection applications that may benefit from the silicon microdosimeters simplicity and portability. Orlic [50] used a silicon photodiode biased at zero voltage, to give an estimated sensitive thickness of 5 μm , and with a thick polyethylene converter mounted in front of the detector. A spectrum using a ^{252}Cf neutron-gamma source was obtained that distinguished the gamma and neutron microdosimetric spectrum. Kadachi [51] extended this work by comparing the response of a silicon PIN photodiode (sensitive depth $\sim 8 \mu\text{m}$ at zero voltage, 13 mm^2 area) and a tissue equivalent proportional counter (TEPC) of 1 μm simulated diameter. The converter for the PIN diode was made by melting wax onto the diode surface. Measurements on both devices were performed during irradiation by an americium-beryllium neutron source. The dose mean lineal energy, $\bar{y}_D$, was found to be 52.1 keV/ μm using the PIN diode and 67.5 keV/ μm for the TEPC with the large difference attributable to the different sensitive volume size and the large path length of recoil protons in the PIN diode. These differences were also evident in the microdosimetric spectra for similar reasons.

Additional comparisons were later made by Kadachi [52] using a variety of silicon photodiodes (p - n , PIN and pnn) and sensitive depths (1,3,5 and 8 μm). The photodiodes were assembled with polyethylene and A-150 TE plastic converters. The effective quality factor, $\bar{Q}$, for irradiation using an Am-Be source was determined from the lineal energy spectra. For a TEPC, $\bar{Q}$ is given by:

$$\bar{Q} = \int Q_{60}(L)d(L)dL \quad (2.30)$$

where $Q_{60}(L)$ is the quality factor function given in ICRP report 60 [53] and $d(L)$ is the normalized dose distribution. Kadachi assumes that the linear energy transfer L is equal to lineal energy y . For the silicon photodiodes, the computation of $\bar{Q}$ required two types of correction:

$$\overline{Q} = \int Q_{60}(L_s)C_Q(L_s)d(L_s)C_d(L_s)dL_s \quad (2.31)$$

where L is the LET of silicon and $C_Q(L_s)$ is a correction factor for the quality factor due to the difference in LET of water and silicon and $C_d(L_s)$ is the ratio of the fractional relative dose of the converter and silicon. $C_Q(L_s)$ was taken by Kadachi to be approximately given by the ratio of the converter to silicon density. The effective quality factor was calculated as 7.9 for the Am-Be source using the TEPC and 8.1 using the 1 μm photodiode ($p-n$) which is quite a good comparison despite the significant spectral differences previously noted. For other applications such as $\bar{y}_D$ determination, the large area photodiode suffers from the problems identified by Dicello [47], namely, a chord length distribution which differs greatly from the proportional counter due to its large area. Furthermore, the sensitive volume of the device is not accurately defined due to the complexity of the charge collection process that includes diffusion and funnel assisted phenomenon (see section 6.1).

2) Arrayed Diode Devices: In 1983, Bradford [4] had clearly recognized the close correlation between radiation effects on microelectronics and radiobiological microdosimetry in a paper entitled "Microelectronic analogues to radiobiological microdosimetry". Single event upset phenomenon resulting from charge collection at a reverse biased node in the memory cell is analogous to biological cell death. Based on the detection ability of memory devices, due to the single event upset phenomenon, several researchers have pursued the idea of using dynamic random access memory (DRAM) circuits as neutron radiation dosimeters. Davis [54] experimented with the thermal neutron response of 16 Kbit DRAMs covered with ^6Li and ^{10}B converters. Similar experiments were also performed by other researchers using different DRAM devices [55-57]. In all these detection systems, a digital output is read from the DRAM and events are simply counted.

However, Abdel-Kader and McNulty [58] in 1987 proposed that the power supply lines of commercial memory devices may be connected to spectroscopic acquisition systems to record the energy deposited by events in reverse biased junctions within the device. Simplistically, a memory cell consists of two transistors in an on (conducting) state and two in a off state (non-conducting). A non-conducting transistor has a reverse-biased $p-n$ junction which is connected via a conducting transistor to the power or ground line of the memory device. When configured with a spectroscopic system connected to the

power line, the device is essentially a radiation detector having a large array of p-n junction cells.

Irradiation with known energy sources is useful for characterizing the sensitive volume dimensions and charge collection regions [58-60]. This technique was further refined by the development of ion beam induced charge collection (IBICC) first performed on microcircuits by Breese [61] in 1992. Sexton [62] provides an excellent review of this subject. Briefly, a narrow ($<1\text{ }\mu\text{m}$) ion beam is magnetically scanned across the device and the charge collected is recorded along with the spatial position of the beam. The location, area and charge collection efficiency of the sensitive volume may then be determined more accurately than by using the broadbeam spectroscopy method (see section 6.2.3).

Roth and McNulty [63, 64] developed a silicon microdosimeter based on memory devices (993L422, 1 KByte) using the principle of power line spectroscopy. The system was intended for biological and electronic radiation monitoring in spacecraft and avionics. An important difference from previous systems is that it used an array of micro-volumes rather than one large detector. Unfortunately, the system was never flight-tested. Stassinopoulos [65] has recently developed a similar system that is undergoing flight-testing. The system uses a $32\text{K} \times 8$ SRAM with no applied bias and an estimated sensitive volume depth of $14\text{ }\mu\text{m}$.

In 1994, Schroder [66-69] applied commercial memory devices to radiation protection applications such as personal neutron dosimetry. Schroder mainly selected radiation hardened memory devices (such as Harris HS6504RH, 4 Kbit) for two important reasons. Firstly, these devices provide designs that are more resistant to radiation damage such as oxide charge build-up (see section 6.3). Secondly, the sensitive volume depth is limited by a high p-doped region below the p-n junction that provides a potential barrier for minority carriers and limits the charge collection from deeper in the substrate. Schroder [68] tested a 4 Kbit radiation hard device with a 2 mm A150 TE converter separated from the device by a 2 mm air-gap and a 64 Kbit non-radiation hard device with a $100\text{ }\mu\text{m}$ thick polyamide layer directly coating the device. The sensitive depth and overlayer thickness were estimated using spectroscopy methods as ($1.9\text{ }\mu\text{m}$, $3.5\text{ }\mu\text{m}$) for the 4 Kbit device [66] and ($3.4\text{ }\mu\text{m}$, $2\text{ }\mu\text{m}$) for the 64 Kbit device [68]. Comparisons with a $1\text{ }\mu\text{m}$ simulated diameter TEPC in a 5.3 MeV mean energy neutron

field (cyclotron) were poor due to deficiencies in the memory device including an ill-defined sensitive volume, large overlayers that prevented heavy ion recoils from reaching the sensitive volume and little consideration to tissue equivalence corrections. The non-radiation hardened device exhibited spectral shifts during irradiation attributed to radiation damage effects.

Schroder [66] along with Pace [70] recognized the possibility of using charge coupled devices (CCD) for radiation protection microdosimetry. The disadvantages of this device are the need for readout control circuitry, the possibility of multiple hits in a single cell before cell readout, and the poor sensitive volume definition. For radiation protection applications, these problems are not as severe as for other applications such as radiotherapy.

Clearly, the recurring major problem in applying commercial memory devices and photodiodes to microdosimetry is the poorly defined sensitive volume compounded by the unwillingness of manufacturers to provide important device construction information. Buried wells are not completely effective at preventing charge collection from outside of the depletion region particularly for high LET particles due to the high charge density exceeding the doping density of such wells. As mentioned earlier, diffusion and funneling severely complicate the charge collection process. Schroder [67] attempted to reduce the effect of diffusion by adding a coupling capacitor following the detector that removed the slower time profile associated with diffusive charge collection. This idea is helpful but nevertheless very imprecise.

Schroder [68] concluded his work with a set of design rules for an optimized semiconductor chip to be used for personal neutron dosimetry. The rules are shown below in Table 2.1. These rules are not exhaustive and not thoroughly justified in the Schroder's paper [68]. This work intends to address the issue of formulating design rules and requirements for silicon microdosimetry.

Table 2.1. Design rules for radiation protection microdosimetry as given by Schroder [68]

Optimization of Design	Criterion
Small diode with $l_{max} / \bar{l} \leq 3$ where l_{max} = maximum chord length and $\bar{l}$ = MCL	Fundamental microdosimetric requirement
Array of diodes with area of a few cm^2 , radiation hardened SiO_2 passivation layer	Efficiency
Minimized SiO_2 thickness, simplified electronics	Threshold
TE converter on top of the chip	Tissue Equivalence

Finally, the concept of simultaneous macro and microdosimetry using a MOSFET was introduced by Rosenfeld in 1996 [71]. For macrodosimetry, the total dose was measured using threshold voltage changes arising from radiation-induced buildup of gate oxide charge. Simultaneously, the reverse-biased drain junction may be used for microdosimetry. Thus, silicon microdosimetry offers the possibility of integrating multiple detector types along with preamplifiers and processing circuitry onto a single small device.

2.3.3 Other methods

For completeness, a few less commonly used experimental microdosimetry methods are briefly described in the following. All the methods, except for microstrip gas counters, are directed at providing more detailed track information than is available using site based detectors such as the TEPC and silicon diode detectors. Recent useful reviews of detector developments are provided by Bartlett [72] and Barthe [73]. These reviews concentrate on neutron personal dosimeters which often use microdosimeter type detectors.

2.3.3.1 Microstrip Gas Counters (MSGC)

MSGCs are similar to multiwire proportional chambers except that the wires have been replaced by alternating anode and cathode strips formed on a supporting substrate. A drift electrode $\sim 1\text{-}4$ mm above the substrate (maintained at the same potential as the cathode) defines the gas volume. These devices are in the early stages of development for microdosimetry having been invented in 1988 by Oed [74]. The principle features of MSGCs of value to experimental microdosimetry are high count rate capability and high spatial resolution [75, 76]. Like the silicon microdosimeter, the tissue equivalence of the device materials (such as substrate) and the effect of the RPP device geometry on microdosimetric spectra need to be evaluated. Thus, some of the work presented in this thesis may have value in the development of MSGCs.

2.3.3.2 Cloud Chambers

Low-pressure cloud chambers have been employed in the study of track structures [77-79]. A three dimensional pattern of droplets created by individual ionizations may be resolved and stereoscopically photographed [79]. Cloud chamber dosimetry provides rich detail on the location of individual ionizations within the gas. However, it is a difficult process generally restricted to the research domain.

2.3.3.3 Optical Ionization Chamber

The three dimensional distribution of individual ionizations with a gas may also be determined using an optical ionization chamber proposed by Turner [80]. The electrons in the particle track are made to oscillate rapidly by the application of an external, short duration, high-voltage electric field. The excited electrons produce additional ionization and electronic excitation of the gas molecules in their immediate vicinity leading to copious light emission (fluorescence) from the gas allowing the location of the electrons to be determined. Two digital cameras perpendicular to one another record the light emission. Turner [80] predicts a resolution capability of ~ 10 nm, based on a position uncertainty of ~ 1 mm at a gas pressure of 5 torr. This may prove a useful tool in the emerging field of nanodosimetry as well as for track structure analysis.

2.3.3.4 Autoradiography

Experimental microdosimetry has been criticized for using detector shapes which may not be representative of the real tissue. This problem is particularly manifest in boron neutron capture therapy where traditional microdosimetry fails to represent realistic tissue architecture and boron distribution. Zamenhof and Solares [81] addressed the issue in the development of high resolution quantitative autoradiography (HRQAR) combined with a novel 2D Monte Carlo simulation. HRQAR uses a $0.8\ \mu\text{m}$ thick film of Lexan as the track detection medium placed in intimate contact with an equally thin tissue sample. Image processing techniques applied to the track detector film following exposure to thermal neutrons identify the boron atoms. Their hypothesis is that a full 3D characterization of the cellular architecture and the corresponding B-10 distribution can be mimicked by sectioning a single plane through many cells in the tumor matrix, deriving the boron distribution within this single plane by HRQAR, and performing a 2D Monte Carlo calculation within this plane over many cell sections to obtain the desired microdosimetric parameters. A spatial resolution of $1\text{--}2\ \mu\text{m}$ and a sensitivity of 1 ppm boron may be achieved, although the process is very time consuming, taking 8 days for the HRQAR. In summary, autoradiography is a specialized time-consuming technique with excellent results for specific applications.

2.3.3.5 Three Dimensional Optical Random Access Memories (3D ORAM)

Three dimensional optical random access memories (3D ORAM) are bistable photochromic materials that can be converted from the stable non-fluorescent form to the quasi-stable fluorescent form (and vice versa) via two photon absorption using

intersecting laser beams. The use of this material for real-time dosimetry was first proposed by Moscovitch [82]. Theoretical calculations by Moscovitch and Emfietzoglou [83, 84] indicated that the energy deposited along a track due to passage of a charged particle through this material results in a local de-excitation of the less stable isomer. The track characteristics depend on the particle type and energy and thus dosimeters based on these materials could provide detailed information on particle type and energy distribution for neutrons and heavy charged particles [84, 85].

Readshaw [86] and Phillips [85] used a photochromic molecule spirobenzopyran (SB) embedded in polymethyl methacrylate (PMMA) to construct track detector films 60 μm thick. The material was read using a confocal laser scanning microscope. Successive two dimensional images formed by scanning across the focal plane (depth profiles) are layered to provide a three dimensional image. Image resolution of the track is approximately 1 μm in the scanning plane and several μm in the depth. Phillips [85] confirmed a fluorescent intensity that is proportional to dose over several orders of magnitude. In the range of energies investigated, the energy per nucleon can be determined from the width of the depth profile. Furthermore, the fluorescent intensity at a fixed dose varies with particle type and energy, a property that may prove useful for particle identification.

3D ORAM is in the early stages of development but promises to be a useful tool for microdosimetry studies of solid-state particle tracks.

2.3.4 Summary of Experimental Microdosimetry Methods

Particle track detection methods such as 3D ORAM, autoradiography, cloud chambers and optical ionization chambers are generally difficult and expensive experimental tools that are operationally time consuming. Site based devices such as the TEPC and the silicon microdosimeter are easier to use, generally applicable and quickly provide microdosimetric spectra and parameters. In this work, we are primarily focussed on developing a silicon based microdosimeter with performance characteristics that are competitive with the TEPC. The ultimate goal is to develop a device that exceeds the usefulness and performance of the TEPC for regional microdosimetry.

Table 2.2. Comparison of proportional counter to silicon microdosimeter

Category	Parameter	Proportional Counter	Silicon microdosimeter
Detector performance	Energy Resolution (Note ii)	Moderate	Moderate,
	Low energy sensitivity (Note iii)	Excellent, Single Ionizations, Minimum $y = 0.05 \text{ keV}/\mu\text{m}$	Moderate, Minimum $y = 0.4 \text{ keV}/\mu\text{m}$
	Sensitive volume definition	Good	Moderate
	Tissue Equivalence	Good	Moderate
	Radiation Hardness	Excellent	Moderate
	Spatial Resolution	Poor, 2.5 cm. 0.5 mm best case [22]	Excellent, 1 μm .
	Wall effect Immunity	Poor	Excellent
	Model cell array	No	Yes
	Shape design flexibility	Moderate	Moderate
Ease of Use	Calibration	Simple	Simple
	Cost	High	Low
	Portability	Moderate	Excellent
	System Complexity	Poor: Requires HV supply and gas supply.	Good: only requires low voltage supply.
	In-vivo use	No	Yes
	Integration	Poor	Excellent

Notes:

- i) The shaded areas identify the best performance of the two devices for each parameter.
- ii) See Figure 2.6 for energy and preamplifier noise dependence. The silicon microdosimeter has potentially better energy resolution (lower $W=3.62 \text{ eVc.f. } 30 \text{ eV}$, better Fano factor and no gas multiplication but needs low preamplifier noise)
- iii) Assuming 2.5 cm spherical TEPC with simulated diameter $d = 1\mu\text{m}$ capable of detecting single ionizations and 1 μm^3 cubic silicon microdosimeter with ultra low preamplifier noise $\sim 15 \text{ erms}$. See section 4.5.6 for discussion of ultimate noise performance of silicon microdosimeter.

A comparison of the typical performance of the TEPC and the silicon microdosimeter is shown in Table 2.2 based on the previous sections discussion. Parameters for comparison are subdivided into "detector performance" criteria and "ease of use" criteria. The best performance for each criterion is shaded gray.

Energy resolution is dependent on preamplifier noise and energy deposited as indicated by equations (2.20) and (2.29). Figure 2.6 shows a resolution comparison of the TEPC and silicon microdosimeter as a function of energy deposited (T) for several preamplifier noise levels. For the TEPC, resolution is dominated by Fano and gas multiplication effects which vary according to $T^{0.5}$ whilst for the silicon microdosimeter preamplifier noise is the dominant factor which varies according to T^{-1} . This accounts for the differing slopes in Figure 2.6.

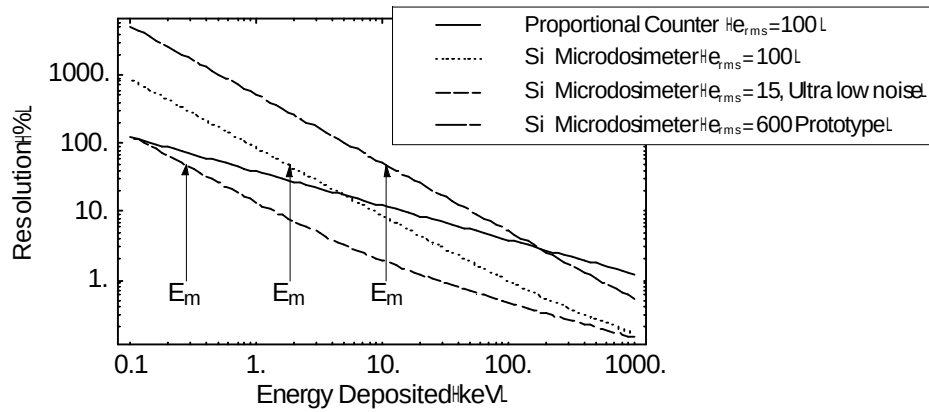


Figure 2.6. Resolution of proportional counter and silicon microdosimeter as a function of energy deposited for several preamplifier noise levels. Calculated using equations (2.20) and (2.29) with $g=10^4$, $M=0.6$, $F=0.3$, $W=30$ eV, $R_{rest}=0$ (TEPC), $F=0.3$, $W=3.62$ eV (Si microdosimeter) Also shown is the minimum energy measurable (E_m lower noise cutoff from equations (2.14) and (2.26)). The E_m for the proportional counter is not shown since it is very low (single ionization).

At higher energies the preamplifier noise is a less significant component of resolution and the theoretical contributions such as the Fano factor and gas multiplication are more significant. Therefore, the silicon microdosimeter has better resolution at higher energies since its theoretical resolution contributions (lower Fano, no gas multiplication) are smaller than the TEPC. The silicon detector's higher preamplifier noise component creates poor resolution at low energies. The minimum energy measurable, E_m , as provided by equations (2.14) and (2.26) is shown in Figure 2.6. The worst case resolution for the silicon detector at E_m is given by substituting equation (2.26) into (2.29). Noting that the Fano contribution is negligible at low energies, we obtain a resolution of 47% at E_m . For an equal preamplifier noise of e_{rms} equal to 100 eV, the silicon microdosimeter resolution is better for deposited energies greater than 4.9 keV. Thus, if a sufficiently low e_{rms} is obtained in the silicon microdosimeter design then the resolution will be better than the proportional counter over most of the measurable energy range. Furthermore, reducing e_{rms} is essential for obtaining an adequate low energy cutoff. Although the potential of the silicon microdosimeter in terms of resolution exceeds the proportional counter we classify both as moderate in Table 2.2 based on current performance as given by the prototype microdosimeter described in this work ($e_{rms}=600$).

The TEPC offers better performance in terms of low energy sensitivity since it is capable of single ionization detection. Sensitive volume definition is good (ignoring the wall effect) and tissue equivalence and radiation hardness are better than the silicon

microdosimeter. For both detectors, the ability to shape the sensitive volume is limited by construction methods. Silicon technology is capable of manufacturing most shapes except for the spherical shape. In all other categories, particularly those relating to ease of use, the silicon detector offers benefits. A higher spatial resolution is especially beneficial in many radiotherapy applications such as proton and heavy ion therapy.

Measurement in the solid phase offers the advantages of removing the wall effect, improving spatial resolution, reducing system size, and portability and offering the ability to model an array of cells. Kliauga [22] in an analysis of the limitations of experimental microdosimetry, recognized the restriction of measurements to the gas phase. He wrote, "Although there is a wide range of opinion as to how serious this constraint is in practice, there is no question that this is a valid issue for development in future research".

Theoretical calculations by Zaider [87, 88] have shown significant phase effects in the estimation of microdosimetric parameters. Calculations of the energy deposition from 30 keV electrons in water vapor and condensed water sites ($d=1\text{ }\mu\text{m}$) gave a 50% higher mean specific energy in condensed water relative to water vapor with commensurately large differences in the lineal energy spectra. Zaider's recent paper [88] stresses the need for development of a solid state microdosimeter suggesting a TE polymer solid state spectrometer as a possible device. To improve signal to noise internal gain could be achieved by operating at a high bias voltage in proportional counter mode rather than as an ionization chamber. Stoner [89] obtained gains of more than 1000 using $7.5\text{ }\mu\text{m}$ thick Kapton foil sheets although recombination is high so that only a maximum of 10% of carriers produced by an ion beam are collected. In this work, we contend that a silicon microdosimeter is a viable alternative (in which close to 100% of carriers produced by an ion beam may be collected), provided very low energy measurements are not required for the applications considered.

The rapidly improving integration capability of silicon technology offers the possibility of reduced system cost, increased reliability and improved performance. Such small systems are capable of in-vivo operation. Moreover, macrodosimetric detectors such as MOSFETs [71] and track structure analysis detectors [90] may be integrated onto the same device to provide high-resolution multi-function capability. However, as a first step towards acceptance a silicon microdosimeter must be developed.

2.4 Applications of Experimental Microdosimetry

The aim of this work is primarily focussed on technology development. Therefore, this section on applications will be a brief introduction. Note that high LET radiotherapy is the primary application pursued in this work. Aside from the discussion below, the various high LET therapy modalities are also discussed in more detail in later chapters.

2.4.1 Radiobiology

The link between experimental microdosimetry, based on regional or site measurements, and radiobiology was forged by Kellerer and Rossi in 1972 with the publication of the Theory of Dual Radiation Action (TDRA) [18]. The theory is based on a number of assumptions, including (from Goodhead [12]):

1. When a biological cell is exposed to ionizing radiation, the radiation produces *sublesions* within the cell and the number of sublesions is directly proportional to the energy imparted by the radiation.
2. A biological *lesion* is formed when two sublesions combine with one another, possibly over relatively long distances on a subcellular scale.
3. Each lesion has a fixed probability of leading to a given biological effect.

Most importantly, the original theory was developed in a site model form compatible with TEPC measurements in simple spherical or cylindrical shapes. This leads to the final simplifying assumption:

4. All pairs of sublesions within a sensitive site have an equal probability of combining and a zero probability outside the site.

Based on these assumptions the TDRA proposes that the mean number of lesions e after an absorbed dose D is:

$$e(D) = k(zD + D^2) \quad (2.32)$$

where k is a constant for a given biological system and z is the dose average of the specific energy (also written as z_D). The first term accounts for lesions formed by two sublesions that are produced in the same event (intra-track action) and the second term pertains to the case of sublesions combining from two separate events (inter-track action).

The relative biological effectiveness (RBE or R) of a high LET radiation H , relative to a low LET radiation L , is defined by:

$$R = \frac{D_L}{D_H} \quad (2.33)$$

where D_L and D_H are the absorbed doses of the two radiations causing equal effect. Utilizing equation (2.32) and equality of effect yields:

$$k(z_H D_H + D_H^2) = k(z_L D_L + D_L^2) \quad (2.34)$$

Substituting (2.34) into (2.33) gives:

$$R = \frac{1}{2D_H} \left[\sqrt{z_L^2 + 4D_H(z_H + D_H)} - z_L \right] \quad (2.35)$$

The site model formulation of TDRA correctly predicts the inverse square root variation of RBE with dose (i.e. $z_L \ll D_H \ll z_H$, $R \approx \sqrt{z_H / D_H}$). The domain in which the specific energy should be considered depends upon the average distance over which sublesions combine. Microdosimetric measurements of the neutron beams employed in biological experiments indicate that the sites typically have diameters of 0.2-2 μm [12, 18].

In the late seventies, it was recognized that the site model formulation of TDRA could not account for the effects of certain radiations such as soft X-rays [91, 92] or correlated pairs of charged particles [93, 94]. For instance, in the molecular ion beam experiment, it was found that if pairs of deuterons are separated by 90 nm the effect on the reproductive capability of V79 cells was significantly different than for a single particle with twice the LET. Yet the microdosimetric spectra for the two kinds of radiation are identical in volumes of the order of a few microns.

The TDRA was generalized by Kellerer and Rossi in 1978 with the introduction of the distance model formulation [95]. The generalized TDRA relaxed the site model assumption (4 above) and replaced it with:

4a. The probability $g(x)$ of sublesions combining is a function of their distance apart x .

According to the distance model z in equation (2.32) becomes (p 242 Rossi [1]):

$$z = \int_0^\infty t(x)g(x)dx \quad (2.36)$$

where $t(x)$ is the proximity function (chapter V Rossi [1]) specifying the spatial properties of the deposition of radiation energy and $g(x)$ is the probability that two energy transfers separated by a distance x combine to form a lesion. The proximity function is a measure of the probability that the energy transfer points are separated by a distance x . On the other hand, the dose averaged specific energy is given by:

$$\bar{z}_D = \int_0^\infty t(x) \left[\frac{f_p(x)}{4\pi x^2 r} \right] dx \quad (2.37)$$

where $f_p(x)$ is the point pair distance distribution (ppdd) of the site where specific energy is measured. Comparing the above two equations indicates that a measurement of $\bar{z}_D$ yields the parameter z only if the microdosimeter and cell sensitive site have the same ppdd. That is;

$$g(x) = \frac{f_p(x)}{4\pi x^2 r} \quad (2.38)$$

Comparisons of experimentally determined $g(x)$ with the function given by the right-hand side of the above equation indicate that the spherical TEPC is a poor representation of the sensitive site geometry [94]. Finding a practically realizable detector structure that satisfies equation (2.38) may be difficult. However, Zaider and Rossi [96] have shown that any function $g(x)$ may be represented as the algebraic superposition of ppdd's corresponding to spherical objects. Thus, the quantity z may be predicted from microdosimetric spectra measured in a series of spherical sites of different diameters. Implementations of these ideas suggest that 2-3 sites are sufficient. The site size required is very small of the order of tens of nanometres, which has consequently driven a demand for nanodosimetric detectors. Variance-covariance techniques (see section IV.5.2 Rossi [1]) may be used for measurements of $\bar{y}_D$ and $\bar{z}_D$ in such small volumes.

Further improvements to the TDRA include the compound dual radiation action theory (CRDA) developed in an attempt to correlate sublesions and lesions with biological entities such as chromosome breaks and chromosome exchange aberrations (p237, Rossi and Zaider [1]). Note that the validity of the TDRA and CDRA are open to much debate with research actively being pursued in alternative radiobiological models such as Goodhead's repair models [12] and work by Zaider and Katz on track structure theory

[97]. Thus, the application of regional experimental microdosimetry as a predictive tool in radiobiology is questionable and requires further research. Silicon based detectors with their flexible geometrical structures and greater design scope may offer new detectors capable of testing track structure theory as well as providing more traditional site based measurements.

2.4.2 Radiotherapy Applications

Microdosimetric methods are especially useful for high LET radiation therapy due to the requirement of quantitative specification of radiation quality. The high LET modalities in which microdosimetric methods have been applied include fast neutron therapy (FNT), boron neutron capture therapy (BNCT), proton therapy (at and beyond the Bragg peak), heavy ion therapy and negative pi-mesons (pions). The silicon microdosimeter developed in this work is tested in the more common modalities of FNT, BNCT and proton therapy.

To gauge the potential importance of these therapies, Wambersie [98] estimated the potential patient recruitment for high LET radiation therapy as about 1-1.5% of the total population. He assumed that cancer incidence is 25-30% of the population, 50% of patients require radiotherapy and about 10% of radiotherapy patients would derive some benefit from high LET radiation. There are about 30000 new cancer cases per year and per million inhabitants in the developed countries. Thus, in Australia (population ~20 million) about 30000 patients per year may potentially benefit from high LET radiotherapy. At present there are about 17 clinical centers worldwide using FNT [98], 14 using proton therapy [99], 7 using BNCT, 3 using heavy ions [100] and only 2 pion therapy facilities [98].

The benefit of high LET radiation therapy, over traditional radiation therapy, is based on two main considerations:

1. Except for protons, the LET is considerably higher than that of electrons. This is important because of the oxygen effect: At low LET, the sensitivity of aerated cells is 3 times greater than anoxic cells whilst at higher LET the difference is much less and the factor approaches 1 for an $LET > 100 \text{ keV}/\mu\text{m}$ [1]. Tumors are generally poorly oxygenated and thus at higher LETs a greater tumor dose may be delivered for the same normal tissue dose.

2. Except for FNT, the spatial specificity of dose delivery is improved with high LET therapy. The depth-dose profile due to the particle Bragg peak offers improved tumor to tissue absorbed doses. For BNCT, particular specificity may be gained due to the local deposition of high boron concentrations within a tumor.

For low LET radiations, absorbed dose is a good predictor of biological effect since the RBE does not vary significantly between X-rays, γ -rays or electrons in the energy range between about 1 to 50 MeV [98] and does not vary with position within the irradiated tissue. Thus, simple absorbed dose measurements suffice and microdosimetric studies are generally not needed for low LET radiations.

In contrast high LET radiations employing heavier particles (neutrons, protons, heavier ions, negative pi mesons) provide an RBE exceeding unity that may vary significantly with depth and beam parameters such as particle energy. For example the RBE of a fast neutron beam may approach 3 for the dose levels and biological endpoints relevant for therapy [98]. A problem specific to high LET radiations is the characterization of RBE spatial variation for a given beam compared to photons. Radiobiological studies such as intestinal crypt colony assay in mice [101] are typically used for RBE studies. However, radiobiological studies are prone to uncertainty due to their difficulty and the large number of parameters involved. Microdosimetric spectrum measurements are a valuable supplement to radiobiological studies in the characterization of RBE. Variations in microdosimetric spectra can identify regions or conditions in which more detailed radiobiological studies may be necessary. Considerable experimental time may be saved by such a protocol.

Mijnheer [102] calculated that the accuracy with which RBE must be calculated for neutron therapy is quite high ($\pm 3.5\%$ 1σ) and at least as stringent as for conventional therapy because of the steep rise and proximity of the curves relating tumor control and normal tissue complications to absorbed dose. The applied RBE directly influences the effective dose delivered to the patient and thus the accuracy requirement for dose also applies to RBE [98].

Calculation of RBE using equation (2.35) is not easily performed in experimental microdosimetry due to the difficulty in estimating the parameter z as discussed in the previous section. Instead empirical approaches have been developed based on calculating parameters from the microdosimetric spectrum $d(y)$ that correlate with RBE.

The parameter $\bar{y}_D$ is a measure of biological effectiveness (i.e. $\bar{y}_D \propto R$) under the conditions of low dose (quadratic term neglected in equation (2.32)), and minimal saturation effects. In an effort to improve the relationship between $\bar{y}_d$ and observed biological effects, Kellerer and Rossi [18] proposed a saturation correction to account for the excessive ionization density of high lineal energy particles over and above requirements for cell death. The saturation corrected dose mean lineal energy y^* is given by:

$$y^* = \frac{\int_0^{\infty} y_{sat} y f(y) dy}{\int_0^{\infty} y f(y) dy} \quad (2.39)$$

where the saturation correction function y_{sat} is:

$$y_{sat} = \frac{y_0^2}{y} (1 - e^{-(y/y_0)^2}) \quad (2.40)$$

with $y_0 = 125$ keV/ μm . Several studies in FNT beams have demonstrated that RBE values calculated from the saturation corrected dose mean lineal energy y^* are fairly well correlated with RBE data derived from radiobiological data [103-105]. Microdosimetric spectra have been extensively used to investigate radiation quality of neutron therapy facilities [106-108]. Comparing the clinical performance of various facilities requires knowledge of the RBE of beams under consideration. Several researchers have compared the radiation quality of neutron therapy facilities using microdosimetric measurements [109-111]. These results have shown that microdosimetric measurements, despite their empirical nature, are capable of discerning even subtle changes in the radiation field and provide insight into the interpretation of such changes.

More references on the application of microdosimetry to FNT, proton therapy, heavy ion therapy and pion therapy may be found in ICRU report 36 (p69 [11]) and Rossi and Zaiders textbook [1] and these will not be repeated here. The specific role of microdosimetry in BNCT, FNT and proton therapy is discussed in more detail in chapters 7, 8 and 9, respectively.

2.4.3 Radiation Protection Applications

Radiation protection applications are primarily concerned with estimating the dose equivalent (H). For particles with a range of LET (L) the dose equivalent is defined as:

$$H = \int_0^{\infty} Q(L)D(L)dL \quad (2.41)$$

where $D(L)=Dd(L)$ is the dose distribution as a function of LET (unnormalized version of $d(L)$) and $Q(L)$ is a quality factor which weights the dose according to biological effect. Various forms of $Q(L)$ have been defined but the standard most recently adopted by the ICRU and ICRP is the definition of $Q(L)$ presented in ICRP publication 60 ([53] and p266 Rossi [1]) and shown in Figure 2.7:

$$Q(L) = \begin{cases} 1 & L < 10 \text{ keV/mm} \\ 0.32L - 2.2 & 10 \leq L \leq 100 \text{ keV/mm} \\ 300L^{-1/2} & L > 100 \text{ keV/mm} \end{cases} \quad (2.42)$$

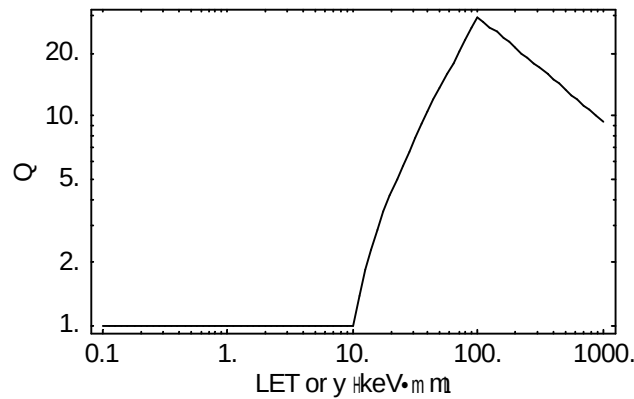


Figure 2.7. Quality factor $Q(L)$ as defined by ICRP publication 60 [53].

Hartmann [112] considered 8 methods of deriving H from microdosimetric data obtained using a TEPC and concluded that all of them had sufficient accuracy for the commonly encountered neutron energy range of between 0.5 and 20 MeV. Note that the accuracy requirements are much less stringent in radiation protection compared to radiotherapy. An overall uncertainty of 30% is considered permissible [113] since many larger uncertainties exist in the application of radiation protection (e.g. position of best measurement, relationship to human body dose). The simplest method of determining H uses the expression $\int Q(y)D(y)dy$ based on the approximation $y \sim L$. Alternatively, Rossi [1] presents a method for determining $D(L)$ from $D(y)$. A more direct assessment of H may then be performed via equation (2.41) and (2.42).

In radiation environments that are essentially unknown or with high energy components, experimental microdosimetry is an important, if not indispensable, tool. However, at low energies in more common environments, factors such as simplicity, cost and portability often favor other approaches. Thus, the employment of microdosimeters in radiation protection has not been widespread despite the possibility of improved accuracy and versatility over current methods. The TEPC has also been known to suffer from aging effects mostly due to chemical gas interactions with material, gas cracking and outgassing of tissue equivalent materials. Despite these disadvantages, portable instruments using TEPCs have been developed, generally employing microprocessors to analyze and control device operation [114, 115].

A silicon based microdosimeter with its construction simplicity, capacity for system integration, and elimination of gas systems offers many benefits in radiation protection applications. However, several fundamental problems inherent in both silicon and TEPC microdosimeter systems must be recognized. At low neutron energies (<100 keV) the short range of recoils can cause substantial errors since the derivation of $D(L)$ from $D(y)$ assumes complete traversal of the sensitive volume. For a TEPC with a $1\text{ }\mu\text{m}$ diameter protons of energy less than 100 keV will not completely traverse the diameter. In addition, microdosimetric measurements in radiation protection are time consuming because of low count rates and may be incapable of measuring even slow changes in radiation quality. Rossi (p271 [1]) states that area monitoring measurements frequently use 10 cm diameter counters simulating $1\text{ }\mu\text{m}$ diameter sites. Typically, counting rates of $500/\text{hr}$ are observed in frequently occupied zones near reactors assuming an ambient dose equivalent rate of around $10\text{ }\mu\text{Sv/hr}$ and an event frequency for fission neutrons of about 0.1 per Gy [1]. Kunz [115] estimated that as few as 16 channels (empirically determined width) are needed in the $d(y)$ spectrum (varying width). Even with this small number of channels, several hours may be required to obtain an adequate spectrum.

2.4.4 Space applications and Radiation Effects on Microelectronics

In a manner analogous to biological cell death, electronic memory cells may undergo state changes if sufficient energy is deposited in the sensitive region of the cell. The sensitive region is localized around the depletion region of reverse-biased p-n junctions. The first evidence of single event upsets (SEU) in memory cells was observed in 1975 by Binder [116] although the effect had been predicted in 1962 by Wallmark and

Marcus [117]. Since then the effect has been extensively studied as testified by the large number of IEEE Nuclear Science journal publications devoted to SEU.

Radiation effects on electronics are more common at high altitudes (avionics) and in space. Moreover, higher radiation levels in space naturally increase the biological risk in manned missions. To study the space environment, TEPCs have been flown on board the space shuttle [118] and Mir space station [119]. The recorded microdosimetric spectra have been used to test radiation transport models and calculations [118]. The development and testing of such models may provide a system for predicting radiation quality during the spacecraft design phase. Microdosimetric measurements have similarly been performed in aircraft in order to quantify radiation effects on personnel from secondary cosmic radiation [120, 121].

Microdosimetric principles and techniques have been applied to several SEU problems based on the previously discussed correlation (see section 2.3.2.3) between SEU and biological cell death (e.g. [122-124] and p291 Rossi [1]). This thesis directly exploits the close link between microdosimetry and SEU by employing a device originally designed to study radiation effects on microelectronics.

2.5 Summary: Key Problems in Silicon Microdosimetry

Based on the review presented in this chapter and summarized earlier in section 2.3.4 we may identify the essential problems impeding the development of silicon microdosimetry.

1. The *requirements* of silicon microdosimetry and particularly the influence of detector *shape* have not been dealt with thoroughly in the literature. This issue will be addressed in chapter 3.
2. The *low noise* requirements on the silicon microdosimeter are demanding and particular attention must be made in the design process to minimize electronic noise. A system noise model and optimization procedure is described in chapter 4.
3. The *tissue equivalence* of silicon microdosimeters has not been addressed in detail in the literature. Chapter 5 provides methods for converting silicon lineal energy spectra into TE spectra.
4. The *sensitive volume boundary* suffers from uncertainty in almost all previous attempts at silicon microdosimetry. Chapter 6 seeks to address the difficult issue of

characterizing the sensitive volume boundary and charge collection efficiency. The use of a parallel array of p-n junction diodes using silicon-on-oxide (SOI) technology, as described in chapter 4, is directed towards improving the definition of the boundary. *Radiation damage* may affect the performance of the silicon microdosimeter by reducing charge collection. This issue is intimately linked to the characterization of charge collection and thus will also be addressed in chapter 6.

If these problems are adequately addressed the many advantages of silicon microdosimetry may prove useful in future and current applications. These include:

- Good energy resolution
- High spatial resolution
- High count rate capability
- Solid-state measurement, wall effect immunity
- In-vivo design
- Manufacturable with good quality control
- Simple, low cost system
- Portable
- Enables integration of multiple detectors, preamplifiers and signal processing components.